EFFECTS OF ULTRASONIC ATTENUATION ON THE ACCURACY OF THE BLOOD FLOW MEASUREMENT TECHNIQUE UTILIZING TIME DOMAIN CORRELATION*

V. T. Suorsa[1,2] and W. D. O'Brien, Jr.[2]

[1]Department of Electrical Engineering, University of Oulu,
SF-90570 Oulu, Finland,
[2]Bioacoustics Research Laboratory, Department of Electrical and
Computer Engineering, University of Illinois, Urbana, IL 61801

ABSTRACT

By utilizing time domain correlation, the volumetric flow in a vessel can be estimated accurately without deteriorating effects due to the frequency-dependent attenuation of ultrasound. When the transmitted pulses are separated by a known time and are reflected by the moving scatterers within the ultrasound beam, the time difference between the pulses changes. The observed time shift is directly related to the radial speed (relative to the beam axis) of the scatterers. If the the flow velocity is sampled at different positions across the vessel, then the volume flow rate can be calculated. The time domain technique has been verified in a blood flow phantom using a blood mimicking substance and natural sponge as attenuating medium between the transducer and the vessel. For hydrodynamically determined flow rates from 50 to 560 ml/min, the continuous volumetric flow has been ultrasonically assessed with an accuracy better than 21%.

I. Introduction

The measurement of human blood flow by Doppler based ultrasound techniques is widely used for diagnosis of vascular disease. Most Doppler flowmeters measure the average flow velocity along the beam direction at a particular distance based on the mean frequency shift [1-3] of the transmitted spectrum. Unfortunately, the Doppler blood flow velocity estimates obtained on the basis of the mean frequency shift can exhibit poor accuracy due to the frequency-dependent attenuation of ultrasound in tissue [4]. The thickness and characteristics of the interposed tissue between the skin and the flow being measured affect the accuracy of Doppler frequency shift estimate. Due to variations in frequency-dependent attenuation [5] and in the thickness of intervening tissue most clinical blood flow measurements are based on qualitative comparisons of measurements from the same patient.

In order to estimate the volumetric flow in a blood vessel both the vessel size and the transducer measurement angle must be known in addition to the flow velocity. The vessel size can be estimated from a knowledge of an average size for the vessel being measured or by using pulse-echo scanning methods [3]. Current Doppler systems, however, do not have any effective means of measuring the transducer measurement angle.

The time domain correlation approach offers several advantages over Doppler based systems in measuring volumetric flow within a vessel. First, the time domain correlation technique calculates the flow velocity from the time it takes a scatterer to move a given distance. This time is directly related to the scatterer speed and it is unaffected by the frequency-dependent attenuation of ultrasound. Second, the vessel dimensions and the transducer measurement angle and finally the volumetric flow can be calculated from the same measurement data.

The time domain correlation technique has been previously demonstrated both on a theoretical basis and on an experimental basis with a volumetric flow phantom [6-9]. The purpose of this paper is twofold: 1) to show results of the time domain correlation technique in a blood vessel phantom with attenuating medium interposed between the vessel and the transducer and 2) to discuss the problems expected to be encountered when applying this technique to humans.

II. Theory

The time domain correlation method tracks scatterers within an ultrasonic beam and determines how far they have moved by the time shift between echoes. Figure 1 shows the geometry of an ultrasonic beam intercepting a vessel. A first echo represents the scatterers inside a range cell illustrated with a solid line. By the time the second echo has been received, the scatterers have moved into the cylinder shown by a dotted line. The time domain technique requires that the time between the initation of each echo (that is the pulse repetition period) is small enough so

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that most of the scatterers stay within the beam and the correlation between the two echoes remains high. As long as this requirement is met, i.e. the dashed cylinder overlaps the solid one in Figure 1, the flow velocity can be calculated as follows

$$V = \frac{c\tau}{2T\cos\theta}$$

where

c = speed of sound in scattering fluid
τ = time shift between the echoes corresponding the maximum correlation
T = pulse repetition period
θ = angle between the sound beam and the velocity vector.

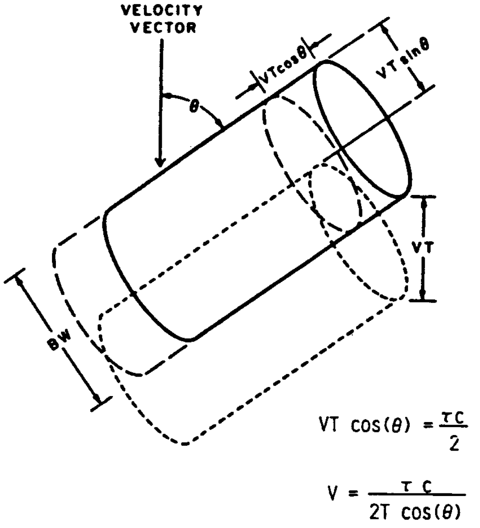


Figure 1. Ultrasonic beam intercepting a vessel [6].

As more scatterers enter the beam between the echoes, the maximum correlation of the consecutive echoes decreases. This is illustrated in Figure 2, which shows the precision of the time domain method. The precision of the time domain method has been previously derived [7,8] and it is defined as the ratio of the standard deviation of the time shift to the mean time shift values. The overall assumption of this derivation is that the signal level is large compared to the noise level. The shown values were obtained experimentally with a 5 MHz transducer and a signal-to-noise ratio of 20 dB [7,8]. For each measurement angle the precision value decreases to a minimum and then begins to increase due to the decorrelation of echoes. In theory, as shown by Embree [7], the precision is expected to decrease by an order of magnitude when the signal-to-noise ratio decreases from 30 dB to 10 dB.

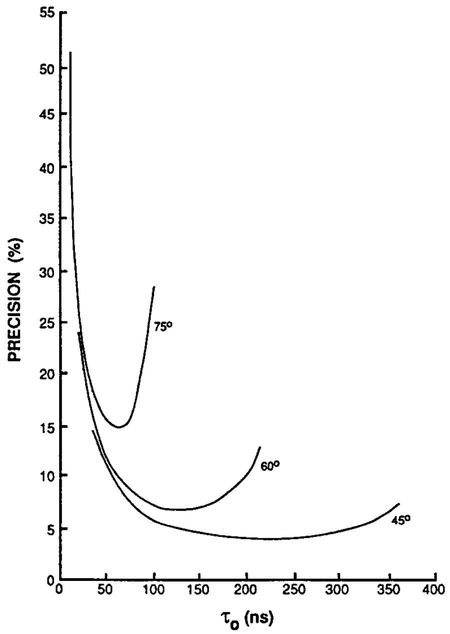


Figure 2. Precision of time domain correlation versus time shift between echoes and measurement angle [8].

The axial flow velocity (i.e. the component of flow velocity in the direction of the ultrasound beam axis) can be determined at different positions across a vessel. As shown in Figure 3, one dimensional flow measurement is defined as the measurement of axial flow velocity versus range along the beam axis. Volumetric flow is estimated from a two dimensional flow measurement. A two dimensional flow measurement is a set of one dimensional measurements at different scanning angles in the same measurement plane, as illustrated in Figure 4. The measurement angle can be determined from a two dimensional measurement by combining the individual one dimensional measurements to a set of constant velocity ellipses, as shown in Figure 5. The average measurement angle is given by

$$\theta_M = \frac{1}{M}\sum_{n=0}^{M-1} \sin^{-1}\frac{a_n}{b_n}$$

where

a_n = half minor axis length of the nth constant velocity ellipse
b_n = half major axis length of the nth constant velocity ellipse
M = number of constant velocity ellipses.

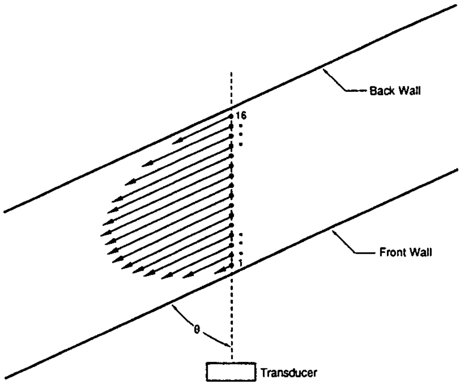


Figure 3. One dimensional flow measurement [9].

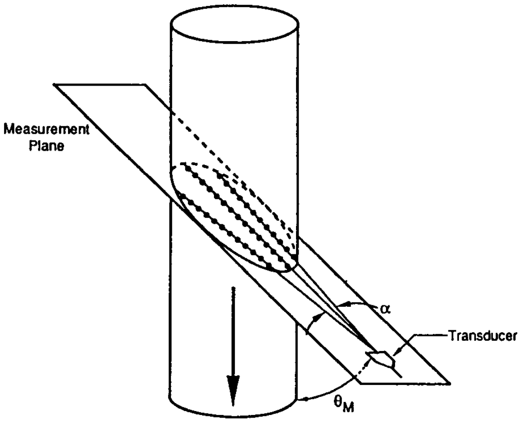


Figure 4. Two dimensional flow measurement [9].

The volumetric flow can be estimated by adding the volume of each M-1 constant velocity ellipse layer. In this case the result is called the INTEGRATED flow estimate. If only the largest constant velocity ellipse is used another volume flow estimate called TUBE SIZE flow estimate is obtained. Thus, in the latter case the entire flow profile must be assumed to be parabolic and axially symmetric while in the former case these assumptions do not necessarily need to be true.

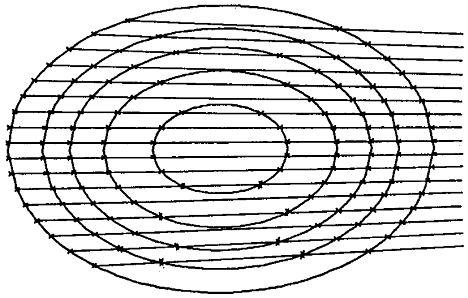


Figure 5. Constant velocity ellipses derived from a two dimensional measurement [9].

To summarize, the effects that are expected due to the introduction of an attenuating medium between the vessel and the transducer is a lower signal-to-noise ratio and thus a degraded precision. Effectively this suggests a worse accuracy for a single or limited set of measurements compared to that of measurements without any attenuating medium.

III. Experimental Methods

To test the effects of scatterers and attenuation on the accuracy of the time domain correlation method the blood vessel phantom illustrated in Figure 6 was used. Dialysis tubing with a 6.3 mm inner diameter was used to mimick the blood vessel and distilled water containing scatterers was pumped through the tubing. A layer of natural sponge of approximately 1 cm thickness was bent around the dialysis tubing and placed between the transducer and tubing. The flow rate was adjusted by controlling the height difference between the upper and lower reservoirs. The volume flow rate was determined by closing the return valve of the lower graduated reservoir and measuring the time to fill the known volume. Beyond a certain length of straight tube, called the entrance length (E in Figure 6), the flow is fully developed and laminar and its velocity profile is parabolic and axially symmetric. The entrance length E for the fluid regulator system is approximately 45 cm and the maximum theoretical value of fully developed flow is about 370 ml/min. In practice, however, the effective entrance length is longer than that shown in Figure 6 and the flow remains parabolic and axially symmetric up to at least 450 ml/min. For flow rates smaller than 100 ml/min a 1.2 m long part of the 6.3 mm diameter plastic tube between the upper reservoir and dialysis tubing was replaced by a 4.7 mm diameter plastic tube of the same length. The large height difference required for this 4.7 mm tube provides better control of low flow rates.

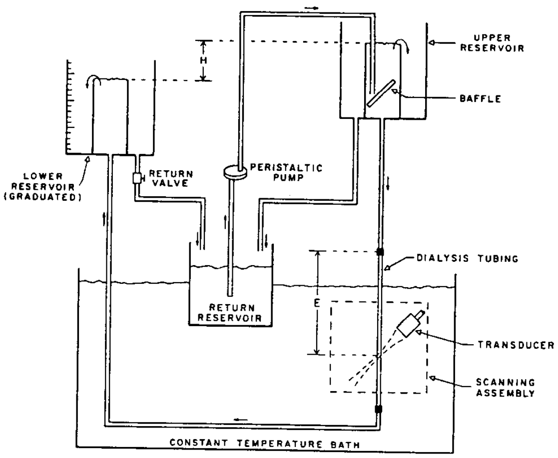


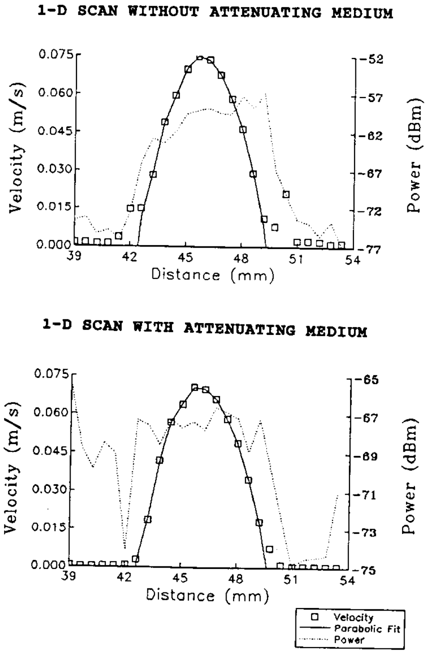
Figure 6. Blood vessel phantom [9].

To obtain a two dimensional velocity field the ultrasound beam was swept across the blood vessel phantom and 11 one dimensional axial flow measurements were performed at the intervals of approximately 0.7°. Three hundred and eighty four echoes were used for each one dimensional measurement. The operation frequency of the transducer (Panametrics model V307) was 5 MHz and its 3dB intensity beamwidth was 2.3 wavelengths (0.7 mm in water) at the focus. The measurement angle (the angle between the ultrasonic beam and the velocity vector) was either 45° or 60°. All the measurements were carried out at the temperature of 23.0 ± 1.5°C.

The received echoes were first amplified and then digitized at the rate of 50 MHz. Next, the data were analyzed using a high speed correlator. Thereafter the data were sent to a Compaq 386/20 personal computer for curve fitting and volumetric flow calculations.

IV. Results

Figure 7 illustrates the differences in one dimensional scans between the measurements made with and without attenuating medium between the transducer and the vessel. The horizontal axis shows the distance from the transducer. The upper curve is a middle scan (α=0 in Figure 4) of a two dimensional measurement without any attenuating medium. The solid line shows the velocity profile after fitting a parabolic curve to the measurement points. The dotted line shows the power of the returned signal. The signal level from the region of the flow is approximately 17 dB greater than that from outside the vessel. The lower curve shows the corresponding data from the middle scan of a measurement with attenuating medium. Here the signal level from the flow is similar to that in front of the vessel. However, the shown signal level is only that in the range gate. In a typical experiment the signal due to attenuating medium was even greater outside the range gate in front of the vessel than in the region of the flow. By setting the measurement angle to 90° and by placing a steel reflector behind the vessel the increase in two-way attenuation due to intervening sponge was measured to be approximately 6 dB. The best precision values of the shown scans are 8.7% for the measurement without attenuating medium, and 9.4% for the one with attenuating medium.


1-D SCAN WITHOUT ATTENUATING MEDIUM

1-D SCAN WITH ATTENUATING MEDIUM

Figure 7. One dimensional scans of velocity and power as a function of distance from the transducer. Upper curve: without attenuating medium. Lower curve: with attenuating medium.

Table I shows the average errors of the time domain volume flow estimates referenced to hydrodynamically determined flow rates. The average errors range from -8.4 to 21% for tube size flow estimates and from -11 to 0.3% for integrated flow estimates. The error percentage remains approximately on the same level throughout the hydrodynamic flow range used in the measurements. The general trend suggested by the results is that the flow estimate becomes more negatively biased due to the attenuating medium.

TABLE I

AVERAGE ERRORS OF VOLUMETRIC FLOW ESTIMATES REFERENCED TO HYDRODYNAMIC FLOW. N IS THE NUMBER OF TRIALS.

	Angle 45°	Angle 60°
A) Integrated Flow Estimates		
Without attenuating medium	0.3% (N=23)	-6.5% (N=19)
With attenuating medium	-10% (N=21)	-11% (N=19)
B) Tube Size Flow Estimates		
Without attenuating medium	21% (N=23)	8.4% (N=19)
With attenuating medium	-6.7% (N=21)	-8.4% (N=19)

Table II summarizes the linear regression analysis results of time domain volumetric flow estimates against hydrodynamically determined volumetric flow for the experiments with attenuating medium. The values for the correlation coefficient range from 0.94 to 0.98 for both volumetric flow estimates. The slope estimates vary from 0.98 to 1.00. As the measurement angle increases from 45° to 60°, the intercept estimates of -8 and -9 ml/min decrease to -20 and -26 ml/min, respectively.

TABLE II

LINEAR REGRESSION ANALYSIS SUMMARY OF COMPARISON BETWEEN TIME DOMAIN AND HYDRODYNAMIC FLOW WITH ATTENUATING MEDIUM.

	Angle 45°	Angle 60°
A) Integrated Flow Estimates		
Hydrodynamic flow range (ml/min)	46-444	72-564
Number of trials	21	19
Correlation coefficient	0.94	0.97
Intercept (ml/min)	-9	-26
Slope	0.98	1.00
B) Tube Size Flow Estimates		
Hydrodynamic flow range (ml/min)	46-444	72-564
Number of trials	21	19
Correlation coefficient	0.95	0.98
Intercept (ml/min)	-8	-20
Slope	0.99	0.99

Table III shows the corresponding data for the experiments without any attenuating medium. Here the values for the correlation coefficient vary from 0.97 to 0.99 for both volumetric flow estimates. The slope estimates range from 0.92 to 0.97 for the integrated estimates and from 1.03 to 1.09 for the tube size estimates. The intercept estimates are positive and less than 10 ml/min except for the tube size estimates with the measurement angle of 45°, where the intercept estimate is 30 ml/min.

TABLE III

LINEAR REGRESSION ANALYSIS SUMMARY OF COMPARISON BETWEEN TIME DOMAIN AND HYDRODYNAMIC FLOW WITHOUT ATTENUATING MEDIUM.

	Angle 45°	Angle 60°
A) Integrated Flow Estimates		
Hydrodynamic flow range (ml/min)	62-454	47-564
Number of trials	23	19
Correlation coefficient	0.98	0.99
Intercept (ml/min)	9	5
Slope	0.94	0.92
B) Tube Size Flow Estimates		
Hydrodynamic flow range (ml/min)	62-454	47-564
Number of trials	23	19
Correlation coefficient	0.98	0.97
Intercept (ml/min)	30	1
Slope	1.03	1.09

The results due to attenuating medium between the transducer and the vessel suggest the following: 1) the slope estimates are closer to unity for both flow estimates indicating that the flow estimates linearily track the actual flow over the extended flow range and 2) the intercept estimates become more negative compared to those without attenuating medium indicating that the flow estimates become negatively biased.

V. Discussion and Conclusions

We have shown that it is possible to measure the volumetric flow by the time domain correlation method, when there is attenuating medium present between the transducer and the vessel. The results suggest that the volumetric flow estimates might become negatively biased because of the attenuation of ultrasound in intervening medium. However, the observed bias is relatively small (ranges from -6.7 to -11%) and the number of trials was only moderate (about 20 for each data set). Further development of our equipment that is presently going on will allow us to make more rapid measurements which will allow us to investigate, whether there really exists a negative bias. In order to use the time domain technique in vivo, the next logical step would be to use attenuating medium between the transducer and the vessel when pulsatile flow is measured. These measurements, together with the further development of our equipment, will allow us to proceed to in vivo experiments.

VI. Acknowledgements

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VII. References

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